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# Effect of Implant–Abutment Connection Design on Load Bearing Capacity and Failure Mode of Implants

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## Abstract

**Purpose:** In this in vitro study, six implant–abutment connection designs were compared and evaluated regarding load bearing capacities and failure modes.
**Materials and Methods:** Five implants of Astra Tech, Bego, Camlog, Friadent, Nobel Biocare, and Straumann were separately embedded in stainless steel tubes using polyurethane, for a total of 30 specimens. Specimens were statically loaded under an angle of 30° with respect to the implant axis in a universal testing machine using a test setup according to ISO 14801. Failure was indicated by a load drop of 100 N in force. Load–displacement curves were analyzed, and maximum force and force at which permanent deformation occurred were recorded. Statistical analysis was performed using one-way ANOVA with the level of significance set at 0.05.
**Results:** Statistical analysis revealed that the type of implant–abutment connection design has a significant influence on load bearing capacity ($p < 0.001$). The mean maximum forces ranged between 606 N (Straumann) and 1129 N (Bego); the forces where plastic deformation set in ranged between 368 N (Friadent) and 955 N (Bego). Failure modes differed between the various implant–abutment connection types tested.
**Conclusions:** Implant–abutment connection design has a significant influence on load bearing capacity and failure mode of implants; however, all implant–abutment connection designs tested would be expected to withstand clinically relevant forces.

In the 1950s, Brånemark et al showed that titanium, experimentally implanted into dogs, was treated as endogenous tissue by the surrounding bone.[1] This phenomenon was named osseointegration.[2] Since then, osseointegrated implants have become increasingly important in dentistry. They are used in a wide range of cases for supporting fixed and removable prostheses.[3]

Within the same basic setup, manufacturers have developed various implant–abutment connection designs. These interface designs can be roughly divided into two groups. The first group may be described as butt joints or slip fit joints (Fig 1A), with a passive connection and a slight space between implant and abutment.[4] The second group comprises conical interface designs with friction fit joints (Fig 1B).[5] Both types can be subclassified into internal and external connection types. With the internal connection type, connective parts of the abutment are placed into the implant body. In contrast, an external connection type is observed when connective parts of the abutment enclose an extension of the implant body. The different implant–abutment connection designs can also be classified with respect to the lock against rotation by an index at the implant–abutment interface. An index is useful in transferring the model cast situation to the in vivo situation by avoiding displacement and rotation of abutment in the fixture.

Norton compared the indexed internal conical interface connection of the Astra Tech (AST) system with Brånemark's hex-indexed butt joint connection and found that the internal conical interface exhibited increased resistance to bending moments at the fixture/abutment interface.[6] Möllersten et al also investigated various implant systems with different joint designs and reported that deep joints exhibited better load bearing capacity than connections with a relatively short overlap of implant and abutment.[7] Additionally, various failure modes (i.e., bent

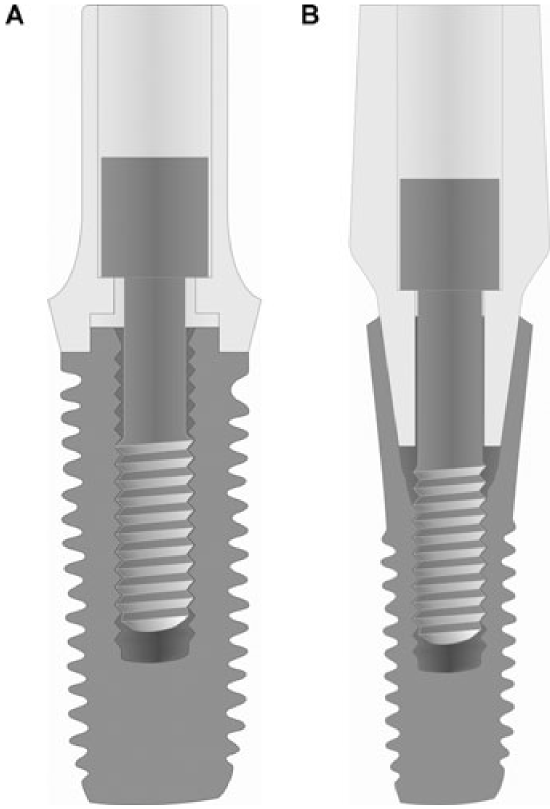


**Figure 1** (A) Example of an external butt joint, (B) Example of an internal friction-fit joint.

or fractured screws or abutments) occurred among the investigated implant systems; however, there was no characteristic failure mode typical for a specific design of implant–abutment connection.

To avoid technical complications during function, one major requirement is a high load-bearing capacity of the implant and its components. A systematic review revealed that technical complications related to implant components and superstructures were reported in 60% to 80% of the studies included, whereas the fixture failed in less than 1% of the cases in vivo.[8] Implant overload was responsible for cracks developing in the material, leading to catastrophic failure even after short periods of function.[9] Failure may also be the result of material fatigue when subcritical slow crack growth develops below the material's yield strength due to cyclic chewing forces. All these complications eventually result in time-consuming and complex treatment, which in the worst case of fixture fracture may end with a large bone defect after explantation.[5]

The aim of this in vitro study was to evaluate the static load bearing capacity of six clinically established implant types with different implant–abutment connection designs. Light-microscopic inspection of fractured specimens was intended to reveal the respective failure modes.

## Materials and methods

Using G-Power (G-Power 3.2.1, Franz Faul, University of Kiel, Kiel, Germany), power and sample sizes were calculated. Power calculation revealed that a sample size of three would have a power of 90% to detect a significant difference in means of 231 N for load bearing capacity (Fm). Power calculation for forces (Fp) at plastic deformation revealed that a sample size of three would have a power of 90% to detect differences in means of 435 N.

Six commercially available implant types (Table 1) were investigated using a static overload test setup according to the standard for fatigue testing of implants and abutments (ISO 14801).[10] Five implants per abutment connection type, with their corresponding abutments and screws, were delivered from commercially available stocks, resulting in 30 specimens investigated in the present study. Each implant was centrally embedded in a polyurethane (PUR, AlphaDie Top, Schütz-Dental, Rosbach, Germany) cylinder, which was framed by a metal sleeve (A2 tool steel) with an inner diameter of 12 mm and a height of 15 mm. The implants were centered in the cylinder with type-specific individual gauges that additionally

**Table 1** Implants, components, and torque used in the current study

| Manufacturer | Implant | Catalogue No. | Abutment | Catalogue No. | Torque (N cm) | Connection type/index |
|---|---|---|---|---|---|---|
| Astra Tech | Osseo speed Ø 4.5/13 mm | 24533 | Ti design 4.5/5.0 Ø 5.5, 1.5 mm | 24235 | 25 | Internal conical interface/hexagon, double hexagon |
| Bego | Semados Ø 4.5/13 mm | 55704 | Sub-Tec Ti abutment S/RI 4.1–4.5 | 56370 | 30 | Internal butt joint with short internal conical matrix/hexagon |
| Camlog | Screw-line promote plus Ø 4.3/13 mm | J1052.4313 | Universal abutment 11 mm | J2211.4300 | 20 | Internal butt joint/3 possible positions |
| Friadent | Ankylos plus B14 Ø 4.5/14 mm | 31021625 | Balance posterior 0.75 | 31021625 | 15 | Internal conical interface/no index |
| Nobel Biocare | MK III groovy RP Ø 4.0/13 mm | 32129 | Easy abutment Bmk syst Rp 1 mm | 30674 | 35 | External butt joint/hexagon |
| Straumann | Standard implant Ø 4.1 RN/14 mm | 043.034S | RN synOcta Ti abutment H 5.5 mm | 048.605 | 35 | Internal conical interface/octagon |

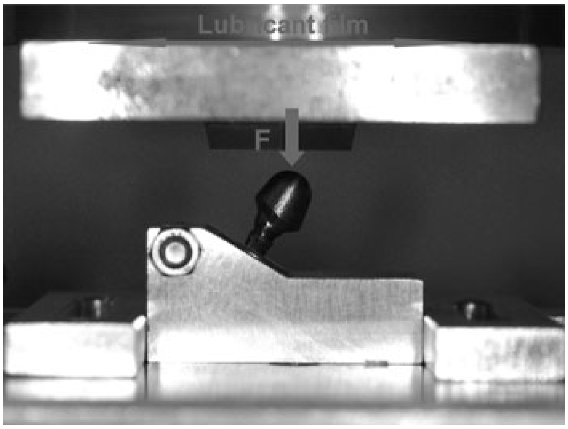


**Figure 2** Astra Tech specimen in universal test instrument prior to loading.

guaranteed simulated bone loss of 3 mm from the implant platform. Afterwards, all abutments were placed on the corresponding embedded implants, and the screws were tightened according to the manufacturers' recommendations (Table 1). A hemispherical loading device made of cobalt chromium alloy (coron, Etkon AG, Gräfeling, Germany) was manufactured by means of computer-aided design/computer-aided manufacturing (Etkon, Etkon AG) and seated onto the unmodified abutments. The distance from the center of the hemisphere to the top face of the PUR cylinder, representing the bone level, was standardized at 11 mm.[10] The specimens were then placed in a stainless steel jig with a 30° angle between the implant axis and the direction of loading. The load test (Fig 2) was performed using a universal testing machine (Type 20K, UTS Testsysteme, Ulm-Einsingen, Germany). A 5 N preload was applied prior to load until failure to ensure that the test specimens were correctly seated in the jig (Fig 2). The off-axis load was applied via a stiff loading plate, which was adhesively connected to the crosshead of the testing machine by a lubricant film (Fig 2). This procedure prevented the loading device from exerting horizontal forces on the implant. The crosshead, moving at a constant speed of 1 mm/min, stopped when failure occurred. Failure was considered to occur when a 100 N load drop was recorded. Load–displacement curves were determined for each implant–abutment connection type and analyzed. In all cases the maximum force, Fm, before failure was regarded as load–bearing capacity. Additionally, the presumed onset of notable plastic deformation was determined for each implant system. For this purpose, the load–displacement curves were fitted by regression lines in the interval between 75 N and 225 N. The force, Fp, at which the load–displacement curve first deviated by 10% from the regression line was recorded as an indicator for initiation of plastic deformation (Fig 3). One-way ANOVA was performed with the level of significance chosen at $p = 0.05$ (SPSS 16.0, SPSS Software Corp., Chicago, IL). A direct comparison of group means (n = 5) was carried out using the post hoc Tamhane test.

After testing, each specimen was embedded in clear methylmethacrylate (Acryfix, Struers GmbH, Willich, Germany) and mid-sectioned along the longitudinal axis by means of a diamond saw (Microslice 2, Metals Research Ltd., Royston, UK). The internal configuration was visually inspected and photographed under a reflected-light microscope (M3Z, Wild, Heerbrugg, Switzerland) at 10× magnification to evaluate the failure mode.

## Results

The load bearing capacity results (Fig 4, Table 2) for Straumann (STR), Friadent (FRI), and AST were significantly lower than for Camlog (CAM) and Bego (BEG). There was no statistically significant difference between the results of Nobel Biocare (NOB) and those of all other groups. The forces at the point when plastic deformation set in, (Fig 5, Table 3) of STR, FRI, and AST were significantly lower than those of CAM. Only the forces for BEG and NOB did not significantly differ from those of the other implant systems investigated.

In the STR group, the internal conical connection of screw and abutment was distorted, and the abutment was dislocated upwards (Fig 6F). The abutment screw always remained nearly intact, whereas the implant neck was bent upwards. With the FRI group, all screw bolts were displaced against their threaded

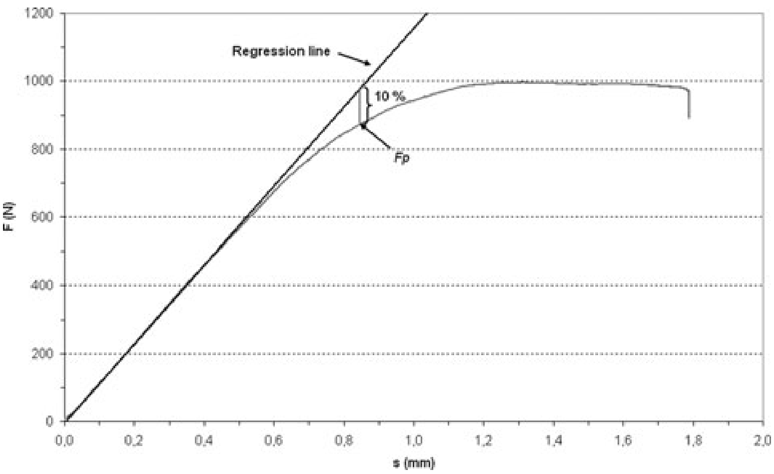


**Figure 3** Load–displacement curve of a Camlog specimen. The force, Fp, at which the load–displacement curve first deviated by 10% from the regression line was recorded as indicator for plastic deformation.

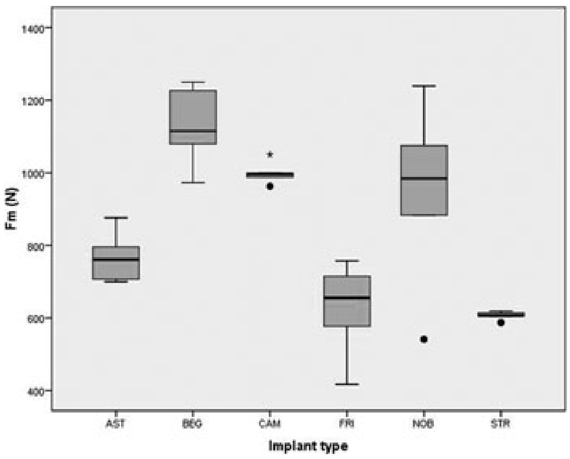


**Figure 4** Load bearing capacity (Fm) versus implant–abutment connection type. Medians, quartiles, extremes, and outliners are given.

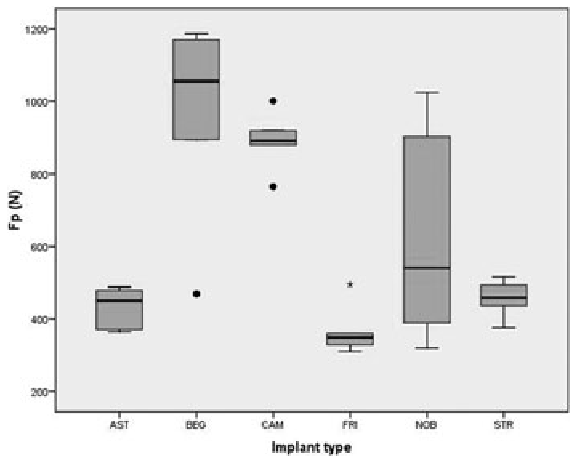


**Figure 5** Force at plastic deformation (Fp) versus implant–abutment connection type. Medians, quartiles, extremes, and outliners are given.

bushings (Fig 6D). Furthermore, the bolts showed considerable necking in the area of their entry into the abutment; in one case, the screw even fractured. While the abutment stayed almost intact, the implant body exhibited fairly large deformations, accompanied by gap formation between abutment and implant body. For AST, the inspection of cross-section revealed only a slight deformation of the abutment, the abutment screw, and the implant body (Fig 6A). There was a gap in upper area of the conical connection between implant and abutment. In contrast, the lower portion of implant–abutment interface still showed tight contact with no observable gap. The abutments of the NOB system showed only a very slight deformation, whereas the implant body was fractured in the region of the butt joint connection in all cases (Fig 6E). Additionally, the abutment screw was deflected, and the abutment was clearly dislocated. Minimal deformation of abutment and abutment screw was observed in the CAM specimens (Fig 6C); however, with the exception of one specimen, all implant bodies fractured between the third and fourth outer thread on the side of load application. After load-to-fracture testing, the abutments of the BEG group were considerably dislocated (Fig 6B). There was an obvious gap between implant neck and abutment. Moreover, the abutment screw was always distorted.

## Discussion

Fatigue testing is accepted as the best way to generate data on fracture strength and longevity of implants and to simulate in vivo conditions.[11-13] Nevertheless, a simple overload test also offers the possibility of drawing conclusions about critical regions of the implant–abutment assembly. Apart from chewing simulation, testing conditions were chosen carefully in the present study to imitate an unfavorable clinical situation with reduced bone support. The implants were embedded with reinforced PUR with elastic modulus similar to natural bone.[14] A distance of 3 mm between implant shoulder and crestal bone level was adjusted to provide a representative case with respect to bone loss.[10] Moreover, in numerous clinical situations, the implant is angulated to the restoration axis.[15,16] Hence, the load was applied 30° off-axis according to the standard for fatigue testing of implants and abutments (ISO 14801) and previous studies.[9,10,17,18] Even though tests were performed under highly realistic conditions, the significance of the present study may be limited due to the sample size of only five specimens per group. Furthermore, just six implant types were evaluated. To achieve a more significant conclusion, implant types with other implant–abutment connection designs have to be evaluated.

Maximum bite forces range approximately between 150 N and 880 N in the posterior region, depending on experimental

**Table 2** Load bearing capacity for the different implant systems

| Implant type | Mean (N) | Standard deviation (N) | Median (N) |
|---|---|---|---|
| AST | 768$^{b}$ | 72 | 761 |
| BEG | 1129$^{a}$ | 113 | 1115 |
| CAM | 999$^{a}$ | 32 | 996 |
| FRI | 624$^{b}$ | 135 | 656 |
| NOB | 944$^{a,b}$ | 261 | 984 |
| STR | 606$^{b}$ | 12 | 607 |

Means, standard deviations, and medians are given. Values denoted by the same superscripts do not differ with statistical significance.

**Table 3** Forces deemed to indicate onset of plastic deformation for the different implant systems

| Implant type | Mean (N) | Standard deviation (N) | Median (N) |
|---|---|---|---|
| AST | 430$^{b}$ | 59 | 451 |
| BEG | 955$^{a,b}$ | 296 | 1056 |
| CAM | 891$^{a}$ | 85 | 891 |
| FRI | 368$^{b}$ | 73 | 349 |
| NOB | 635$^{a,b}$ | 313 | 540 |
| STR | 456$^{b}$ | 54 | 459 |

Means, standard deviations, and medians are given. Values denoted by the same superscripts do not differ with statistical significance.

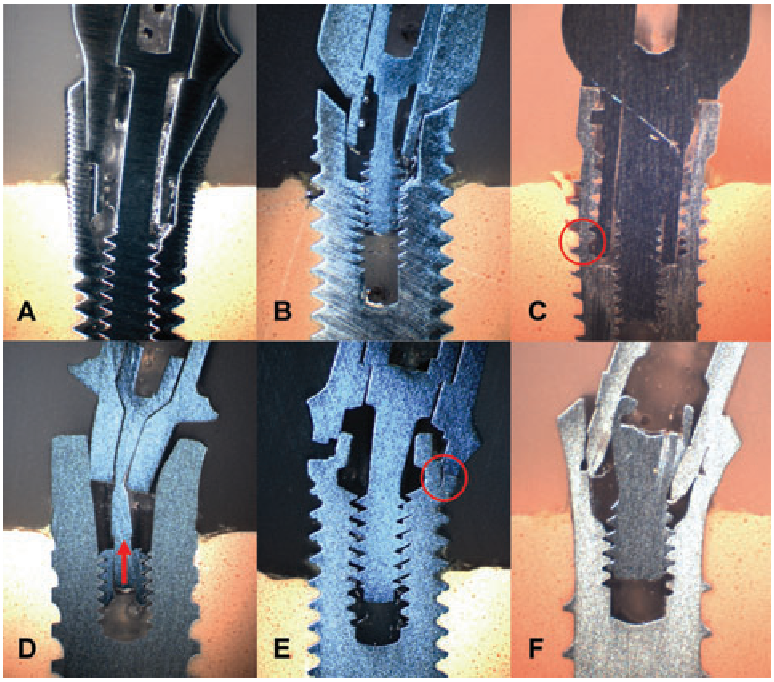


**Figure 6** Polished cross-sections of embedded failed specimens of the different implant–abutment connection types. (A) AST, (B) BEG, (C) CAM (implant body fracture indicated by red circle), (D) FRI (displacement of screw bolt against threaded bushing in direction of arrow), (E) NOB (implant body fracture indicated by red circle), (F) STR.

condition.[19-22] In contrast, average bite forces only range between 20 N and 120 N, depending on food hardness.[23] All implant–abutment connection types tested in the present study showed mean load bearing capacities (Fm) near the above-mentioned maximum bite forces or even above these. These values represent forces where a more or less large permanent deformation of the implant–abutment assembly had already taken place. Hence, the forces at which plastic deformation set in were also reported (Fp). These forces range above average bite forces, but maximum bite forces may reach the level of Fp; however, the testing conditions (3 mm bone loss, 30° off-axis load) were adapted to a clinical worst case. With properly placed implants without crestal bone resorption, the forces expected to cause plastic deformation are usually considerably higher.

With the STR group, the failure of the implant–abutment assembly was not characterized by an abrupt load drop due to the components' fracture, but by a major dislocation of the abutment and the implant body. Visual inspection of the cross-sectional area of the failed implants showed that the critical zone was the head of the abutment screw and its internal conical connection to the abutment. In these units, the internal abutment cone widened, and the abutment nearly slipped over the screw head (Fig 6F). Notwithstanding the low load bearing capacity of the STR system, finite element analysis revealed that this implant system is not prone to fracture.[24] Moreover, the design of the STR system is well documented and has been in use since 1985.[25,26] Buser et al reported only low rates of failure for an observation period up to eight years.[27]

Regarding the FRI group, the abutment screws were found to be weak spots in the system. The connection between screw bolt and threaded sleeve was the first part to fail during loading (Fig 6D). In contrast to all other implant systems tested, which use one-piece abutment screws, the FRI screw was made of two components. This kind of construction probably causes a significant decrease in load-bearing capacity, which could perhaps be avoided by the use of one-piece abutment screws. A further distinctive aspect of the failure mode was the considerable necking in the area of the abutment screw's entry into the abutment. This also may be due to the disconnection of the screw parts, resulting in dislocation of the abutment out of the implant body, accompanied by a loose fit. Hence, the stabilization of the abutment against horizontal forces is reduced, and consequently stresses at the abutment screw may increase. Regardless of the deformations monitored under the relatively high load applied in the present study, the FRI system shows good performance under clinical conditions.[28,29]

Compared to the FRI implants, the AST group, also featuring a conical connection design, showed only slight deformations of the abutment screws (Fig 6A). There may be two reasons for this phenomenon. First, the joining surface of the conical implant–abutment connection is greater with the AST system, which may improve load distribution.[30,31] Second, the diameter of the one-piece AST screw is significantly larger, leading to greater resistance to bending forces. This advantageous design of the implant–abutment connection may be one important reason for the good clinical performance, although other aspects, such as implant surface design, are also important.[32-37]

Within the NOB group, fracture of the implant body was identified in the region of the butt joint connection (Fig 6E). This may be due to the short external hex, which does not stabilize the abutment against loads in horizontal directions,[38] resulting in a deflection of the abutment with only a small supporting point located at the area of the implant shoulder opposite to the side of load application. Hence, not the abutment, but the screw and the implant shoulder, have to withstand most of the load; however, one reason for the comparatively high load-bearing capacities may be the improvements in screw design and the change to a more ductile alloy by the manufacturer.[39,40] Consequently, the implementation of these constructional changes could improve the clinical outcome.[3,38,41–43]

A further mode of failure was observed in the CAM group. The implant wall fractured in almost all cases approximately 5 mm below the implant shoulder, at the side of load application (Fig 6C); in contrast, the complex of abutment and screw showed only a slight deflection. This may be a result of the relatively rigid and massive abutment and abutment screw, which transfer most of the load toward the center of the implant. But particularly that region of the implant where highest stresses occur proved to be a locus minoris resistentiae: the wall thickness of the implant in this area is reduced due to the groove-like depressions of the inner and outer thread giving rise to stress concentrations and subsequent crack development. Nevertheless, the CAM system also shows good treatment outcomes under clinical conditions,[44] and higher load-bearing capacities were investigated than with other systems. The main reason for this positive outcome may be the tube-in-tube connection.[38]

The BEG system also has a kind of tube-in-tube connection, with a small angulated surface in the implant shoulder area. Considerable dislocation of the abutment was observed, with gap formation (Fig 6B); however, the BEG system showed the highest load-bearing capacities in the present study. This may be explained by the massive design of the implant walls, which could thus withstand the applied forces. Therefore, in contrast to the CAM system, the implant was not deformed or even fractured, but the abutment screw was substantially elongated. While the current study gave promising results with respect to the mechanical performance, the authors are unaware of any published studies on the long-term clinical outcome of the BEG implant system.

Statistical analysis revealed significant differences in load bearing capacity between the different implant–abutment connection types investigated in this study (Table 2). Nevertheless, load-bearing capacities with all types were considerably higher than average chewing forces.[23] It should be emphasized that the results of a static overload test of load-bearing capacity do not allow reliable conclusions with respect to long-term clinical success. Performance under functional loading in vivo depends on additional aspects, for example, microgap formation and screw loosening.[45,46]

## Conclusions

Within the limitations of this study, it could be concluded that:

1. Implant–abutment connection design has a significant influence on load-bearing capacity of implants.
2. Failure mode due to static overload differs between implant–abutment connection designs.
3. All of the implant–abutment connection designs tested would be expected to withstand clinically relevant forces.
4. Long opposing lateral surfaces of implant and abutment seem to have advantages with respect to load-bearing capacity in comparison to connections with a relatively short overlap of implant and abutment.

To draw conclusions for clinical long-term behavior, further parameters like cyclic fatigue have to be taken into account in further investigations.

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